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1 **Physical principles of medical ultrasound**

2

3 Michiel Postema^{1,2,3}, Spiros Kotopoulos⁴, Klaus-Vitold Jenderka⁵

4

5 ¹ School of Electrical and Information Engineering, University of the Witwatersrand,
6 South Africa;7 ² Inserm Research Unit U930: Imaging and Brain, Université François-Rabelais de
8 Tours, France;9 ³ LE STUDIUM Loire Valley Institute for Advanced Studies, Orléans, France;10 ⁴ Department of Gastroenterology, Haukeland University Hospital, Bergen, Norway;11 ⁵ Department of Engineering and Physics, Merseburg University of Applied
12 Sciences, Merseburg, Germany.

13

14 **Sound and ultrasound**

15 Acoustics is the scientific field that studies sound. Sound is a form of mechanical
16 periodic molecular displacement (vibration) of matter. The time it takes for a vibration
17 cycle to complete is called a period. The number of vibration cycles that occur during
18 a set time is referred to as the frequency. The frequency f of a vibration is the inverse
19 of its period T :

20

$$f = \frac{1}{T}. \quad \text{Eq. 1}$$

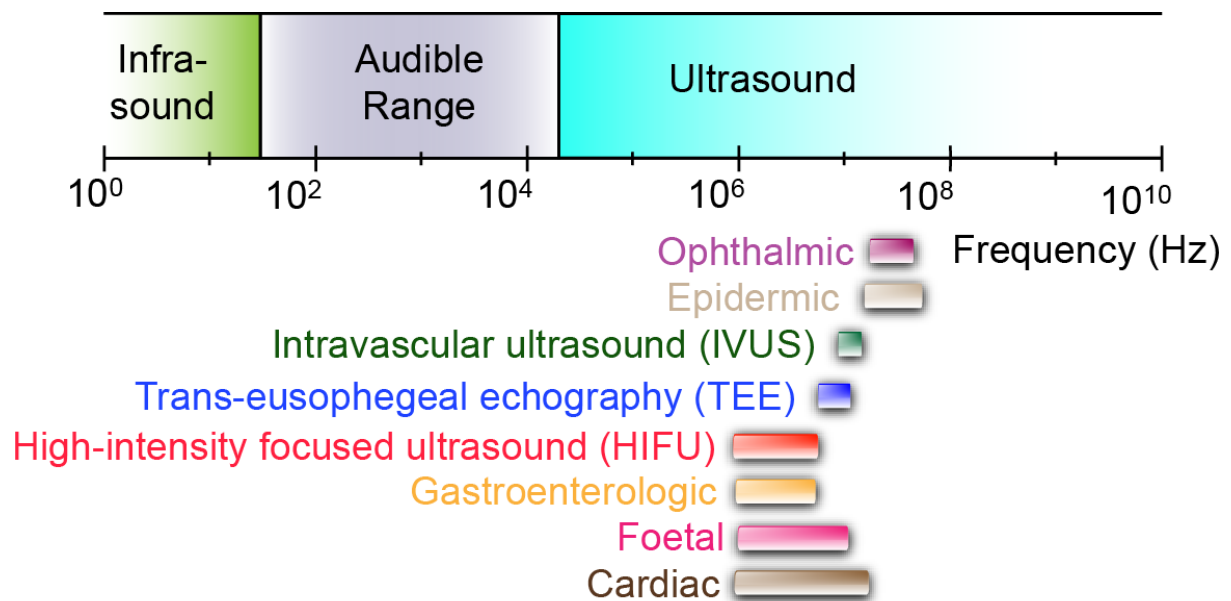
21

22 Sound with frequencies below 20 cycles per second, *i.e.*, below 20 Hz, is called
23 infrasound. Although infrasound is too low to be heard by human beings, it can be
24 perceived (felt).25 The audible range is defined by frequencies between 20 Hz and 20,000 Hz (20 kHz).

26 This range has been defined by the average hearing of healthy 18-years-old men.

27 Frequencies higher than 20 kHz are referred to as ultrasound.28 Figure 1 shows some clinical application of ultrasonics and their respective frequency
29 bands.

30



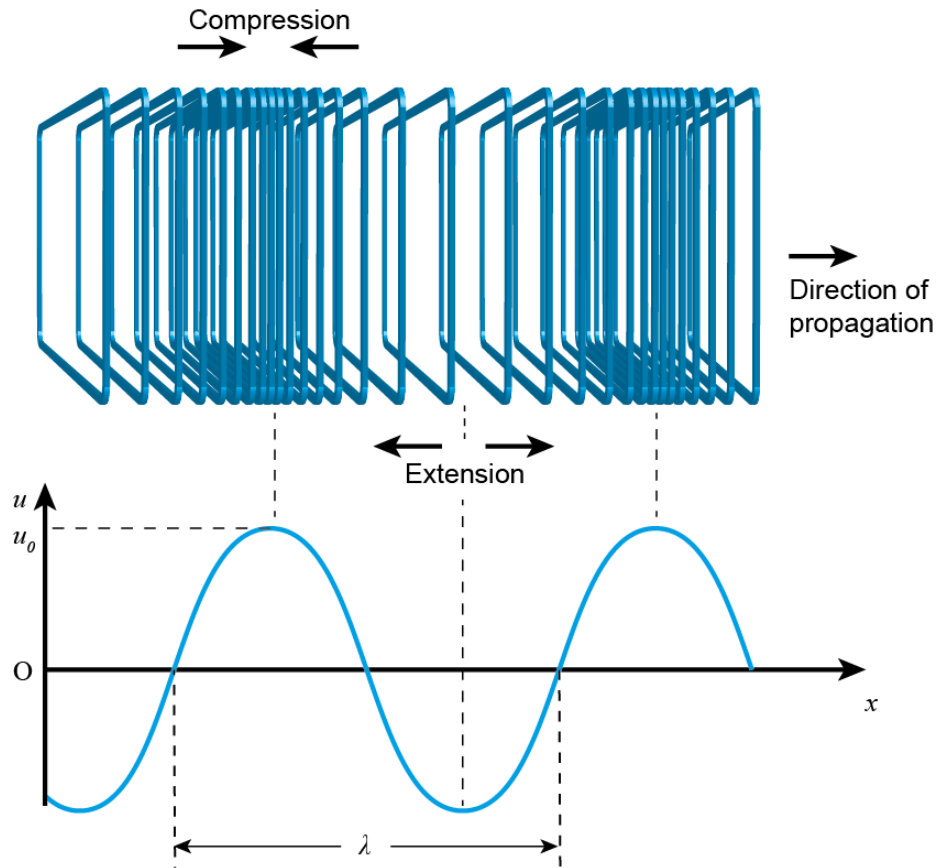
31

32 **Figure 1 – Clinical applications of ultrasound and their corresponding**
 33 **frequency bands.**

34

35 Sound propagates from a source through matter. Although many different elastic
 36 wave types exist in solid materials, fluids only support longitudinal waves.
 37 Longitudinal waves, also known as acoustic waves, displace matter only in the
 38 direction on propagation (*cf.* Figure 2). As human tissue consists mostly of fluid
 39 materials, primarily longitudinal waves are generated and observed in the field of
 40 medical ultrasonics.

41



42

43 **Figure 2 - Schematic representation of the axial displacement of matter by a**
 44 **longitudinal sound wave.**

45

46 The highest displacement in a sound wave is called the displacement amplitude.
 47 Generally, the matter displacement in space and time by a low-amplitude sound
 48 wave has the form

49

$$u(x, t) = u_0 \sin 2\pi \left(\frac{t}{T} - \frac{x}{\lambda} \right), \quad \text{Eq. 2}$$

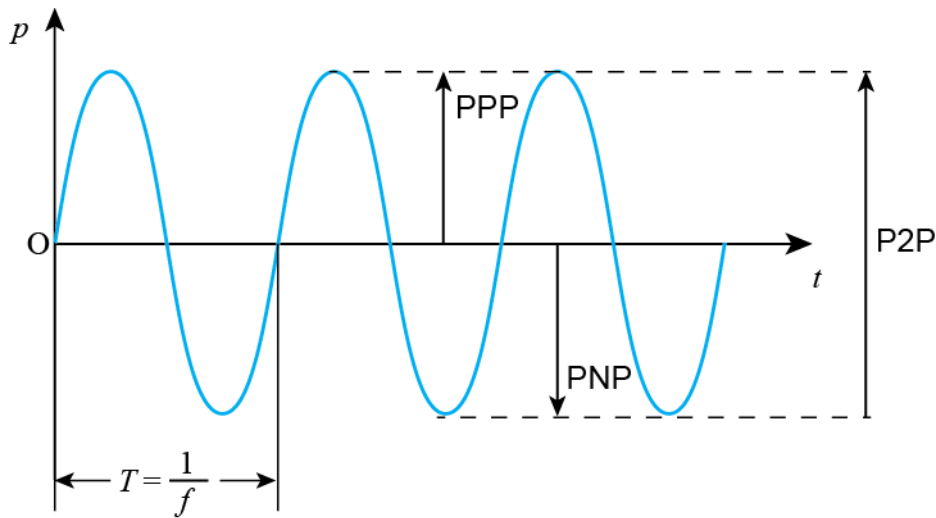
50

51 where u_0 is the displacement amplitude and λ is the wavelength of the sound (*cf.*
 52 Figure 2).

53

54 Notice the minus between $\frac{t}{T}$ and $\frac{x}{\lambda}$ in (2): obviously, the wave at given time farther
 55 from the source is equal to the wave at earlier time closer to the source. Taking only
 56 one dimension into account, the compressive and extensive displacements are
 57 related to local pressure changes by the equation of motion from which the wave
 58 equation is derived (*cf.* Appendix, Eq. A 1 – A 3).

59



60

61 **Figure 3 – Common parameters used to express pressure amplitudes: peak-**
 62 **positive or peak-compression pressure (PPP), peak-negative or**
 63 **peak-rarefaction pressure (PNP), peak-to-peak pressure (P2P).**

64

65 Figure 3 shows some often-used parameters to express the pressure amplitudes of
 66 medical ultrasound, which are handy especially if the sound waves are asymmetric.

67

68 Let us define an imaginary sound source with power W , *i.e.*, every second, a certain
 69 amount of energy is radiated from the source. The power is an intrinsic property. At
 70 equal distances from the source, we can define a surface S , through which this
 71 energy must pass. The power per unit surface area is called the instantaneous
 72 intensity:

$$I = \frac{W}{S}. \quad \text{Eq. 3}$$

73

74 The averaged derived intensity of a harmonic sound wave at a point in a sound field
 75 is:

76

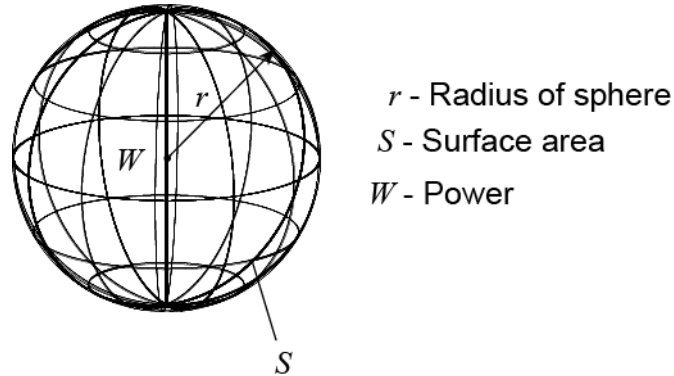
$$\langle I \rangle = \frac{p_A^2}{2\rho c}, \quad \text{Eq. 4}$$

77

78 where p_A is the pressure amplitude, c is the speed of sound in the medium, and ρ is
 79 the density of the medium. Thus, for a point source, the surface through which the
 80 energy must pass is a sphere of radius r (*cf.* Figure 4) and a surface area $S = 4\pi r^2$.
 81 Consequently, for a point source, the intensity is inversely proportional to the

82 distance to the source squared, and the acoustic pressure is inversely proportional to
 83 the distance itself. This acoustic pressure decay with distance is called geometric
 84 damping.

85



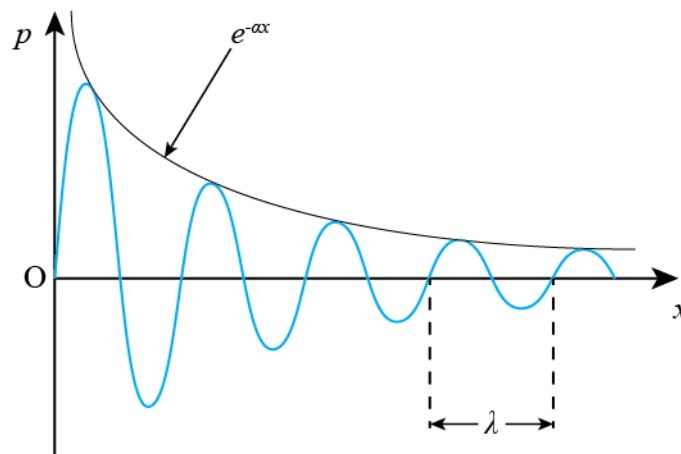
86

87 **Figure 4 – Radiated field through a spherical surface S at a distance r from a**
 88 **point source with power W .**

89

90 Thermal and viscous material properties are other causes of damping of the acoustic
 91 wave (*cf.* Figure 5). Damping coefficients are frequency-dependent. In human tissue,
 92 the damping coefficient is proportional to the frequency to a power between 1.0 and
 93 1.4. Thus, the higher the frequency, the lower the penetration depth of the sound.

94



95

96 **Figure 5 – Damped wave with wavelength λ and damping coefficient α .**

97

98 The amplitude of a received acoustic signal is generally expressed in decibels
 99 relative to a reference pressure:

100

$$\text{SPL} = 10 \log_{10} \left(\frac{p}{p_{\text{ref}}} \right)^2 = 20 \log_{10} p - 20 \log_{10} p_{\text{ref}}, \quad \text{Eq. 5}$$

101

102 where SPL is the sound pressure level in decibels. Decibels are always rounded to
 103 whole numbers. Table 1 gives some typical values for pressure changes and their
 104 respective level in decibels.

105

SPL [dB]	Multiplication
-20	0.10×
-12	0.25×
-6	0.50×
0	1×
6	2×
12	4×
20	10×
40	100×
60	1,000×
80	100,000×

106

Table 1 – Sound pressure levels and their corresponding multipliers.

107

108 Most acoustic waves propagate unhindered through the human body. A small
 109 proportion is specularly reflected on tissue transitions. The amount of reflected sound
 110 at such a boundary is dependent of the acoustic impedances on both sides of the
 111 boundary. The acoustic impedance Z of a medium is defined by

112

$$Z_i = \rho_i c_i, \quad \text{Eq. 6}$$

113

114 where c_i and ρ_i are the speed and the density, respectively, of medium i . Reflection
 115 and transmission coefficients are used to predict reflections from boundaries.

116 In most organs, tissues have rather small acoustic impedance differences. The
 117 boundaries consist of cells with sizes much smaller than the wavelength of the
 118 ultrasound used for imaging. The signals travelling back to the sound source from
 119 tissue transitions are actually caused by scattering. Given the long wavelengths of
 120 the ultrasound, cells can be considered point scatterers. The backscattering from

121 point scatterers is proportional to the number of scatterers per volumetric unit
 122 (scattering density), proportional to the square of the combined compressibility and
 123 density differences of the scatterers, inversely proportional to the fourth power of the
 124 wavelength and therefore proportional to the fourth power of the frequency, and
 125 proportional to the sixth power of the radii of the point scatterers. For larger
 126 scatterers, such as collagens or veins, the scattering behaviour is different from the
 127 so-called Rayleigh scattering from point scatterers. The backscattering properties
 128 have been quantified for many structures of millimetre-size in organs. Using these
 129 quantifications of backscattered signal, abnormalities can be traced. As an example,
 130 fatty liver cirrhosis can be traced from the change in scattering from enlarged mean
 131 distances between lobular structures.

132

133 Moving scatterers such as blood cells create a shift in the ultrasound signal. This so-
 134 called Doppler shift can be approximated by

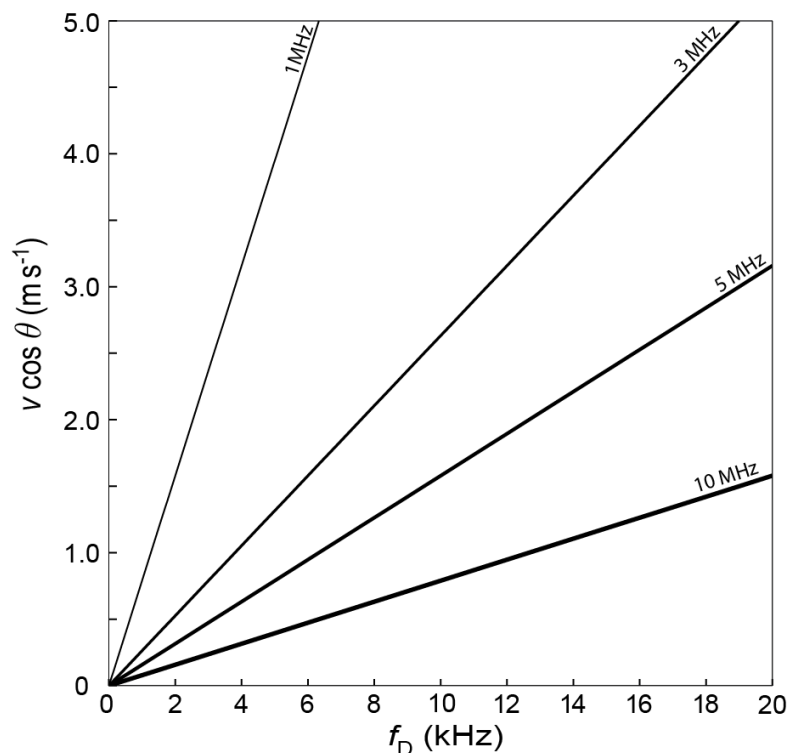
135

$$F_D \approx 2f \frac{v}{c} \cos\theta, \quad \text{Eq. 7}$$

136

137 where θ is the angle between the ultrasound beam and the streaming direction
 138 (positive axis) and v is the magnitude of the streaming velocity (cf. Figure 6).

139



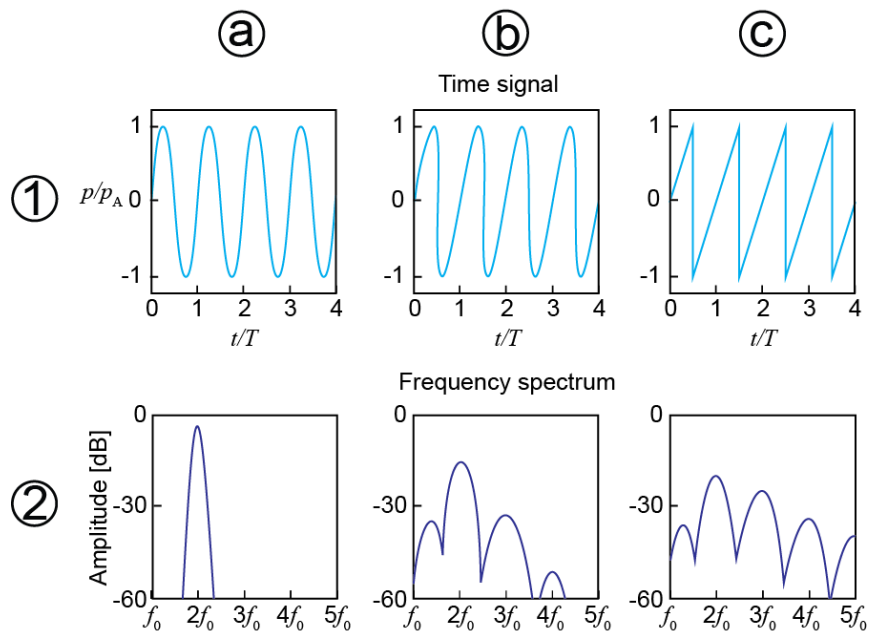
140

141 **Figure 6 – Lateral velocity as a function of Doppler shift at four different**
 142 **transmitting frequencies.**

143

144 Changes in the frequency of the signal are also caused by nonlinear propagation
 145 through tissue and by the presence of ultrasound contrast agents. Nonlinear
 146 propagation is caused by the fact that the speed of sound in compressed tissue is
 147 slightly higher than in extended tissue. Therefore, the peaks of ultrasound waves
 148 travel faster than the troughs. The waves are distorted farther away from the source,
 149 until only saw-tooth shapes remain. Figure 7 shows some waveforms at different
 150 distances from the source, and their frequency content.

151



152

153 **Figure 7 – Waveforms at different distances from the source, and their**
 154 **respective frequency spectra.**

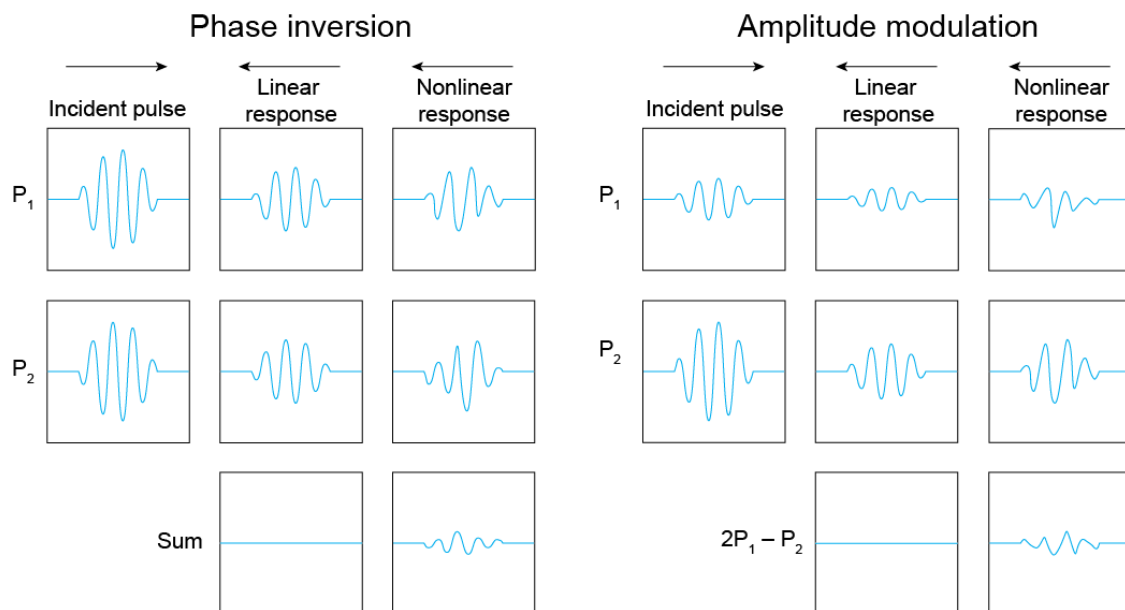
155

156 Blood cells are poor scatterers in diagnostic ultrasound. Because perfusion imaging
 157 is often desired in clinical diagnosis, ultrasound contrast agents have been injected to
 158 enhance the scattering from blood. Ultrasound contrast agents consist of
 159 microscopically small perfluorocarbon gas bubbles encapsulated by elastic (most
 160 commonly phospholipid) shells. These microbubbles oscillate linearly and nonlinearly
 161 in sound fields, radiating a detectable acoustic signal. Several detection strategies
 162 exist to reveal the presence of microbubbles and therefore blood (*cf.* Figure 8).

163 Recently, the peculiar behaviour of microbubbles under specific acoustic conditions
 164 close to living cells has led to research into therapeutic applications of microbubbles

165 whose shells have been modified to contain drugs or genes. Ultrasound-guided drug
 166 delivery might be possible using regular clinical ultrasound equipment.

167



168

169 **Figure 8 – Two detection strategies for the presence of microbubbles.**

170

171 **Transducers**

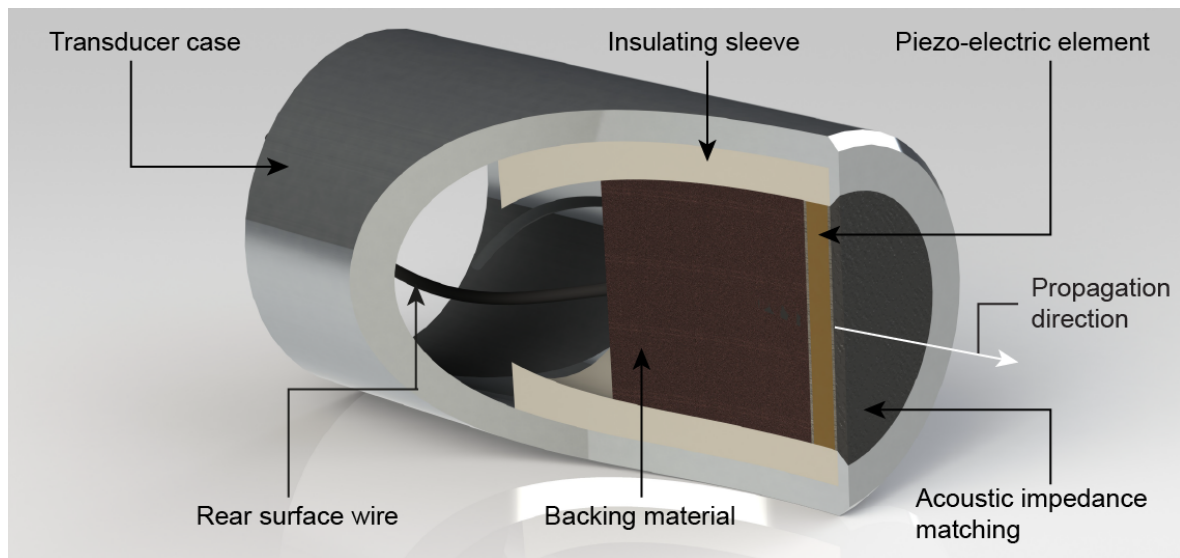
172 Ultrasound transducers convert electrical signal to pressure waves and vice versa.
 173 With therapeutic devices, such as those used for physiotherapy or ultrasound-
 174 mediated surgery, only the transmit capability is used, whereas diagnostic devices
 175 both transmit and receive. In all cases, transducers contain piezoelectric elements to
 176 generate ultrasound or convert ultrasound into an electrical signal.

177 When strain is applied, the electric charges in the elements are redistributed,
 178 therefore generating an electrical impulse. Inversely, when an electrical impulse is
 179 applied, it changes the geometry of the piezoelectric material. This is true for all
 180 piezoelectric materials.

181

182 Apart from one or more piezoelectric elements with electrodes attached to both
 183 sides, ultrasound transducers consist of a backing behind the element, and one or
 184 more matching layers in front of the element (*cf.* Figure 9).

185



186

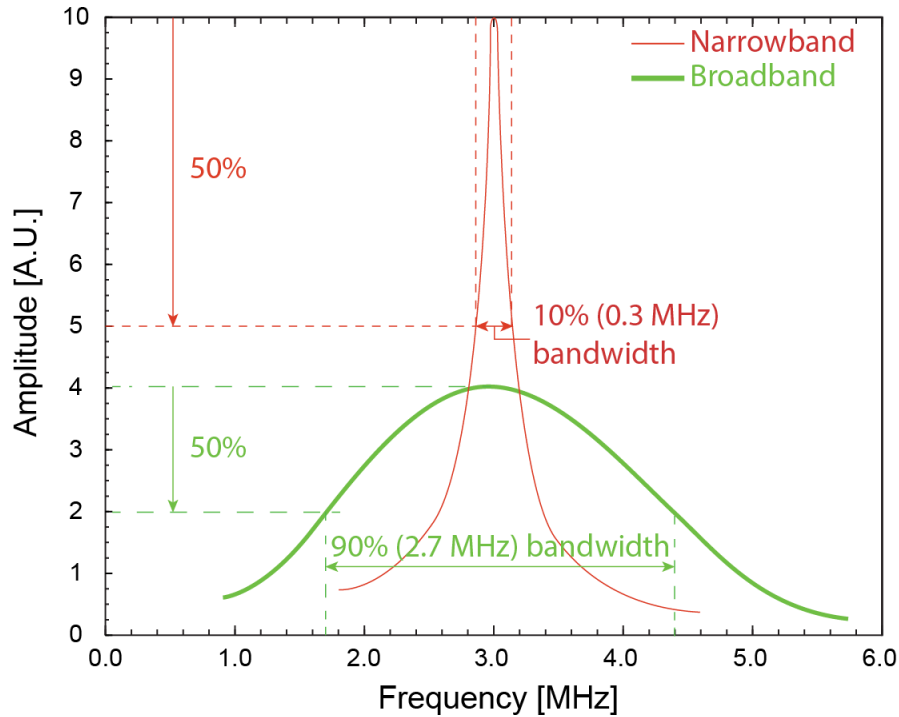
187

Figure 9 – Components of a single-element transducer.

188

189 The thickness of the element determines its resonance frequency: its natural
 190 oscillation frequency. Without backing present, the element can oscillate with
 191 maximum amplitude, *i.e.*, extend and contract, at this frequency. The backing
 192 material determines the bandwidth of the transducers. The bandwidth is the
 193 frequency band at which a transducer generates and receives sound (*cf.* Figure 10).
 194 The choice of backing material is critical for the performance of the transducer. The
 195 matching layer forms a near-lossless transition between the element and the
 196 medium.

197



198

199

Figure 10 – Tradeoff between higher-power output and wide bandwidth.

200

201 A transmitting transducer creates a sound field. Close to the transducer surface,
 202 interference causes local pressure variations (*cf.* Figure 11). The width of this so-
 203 called near field is roughly equal to the diameter of the transducer. Its length is given
 204 by

$$N = \frac{D^2}{4\lambda}, \quad \text{Eq. 8}$$

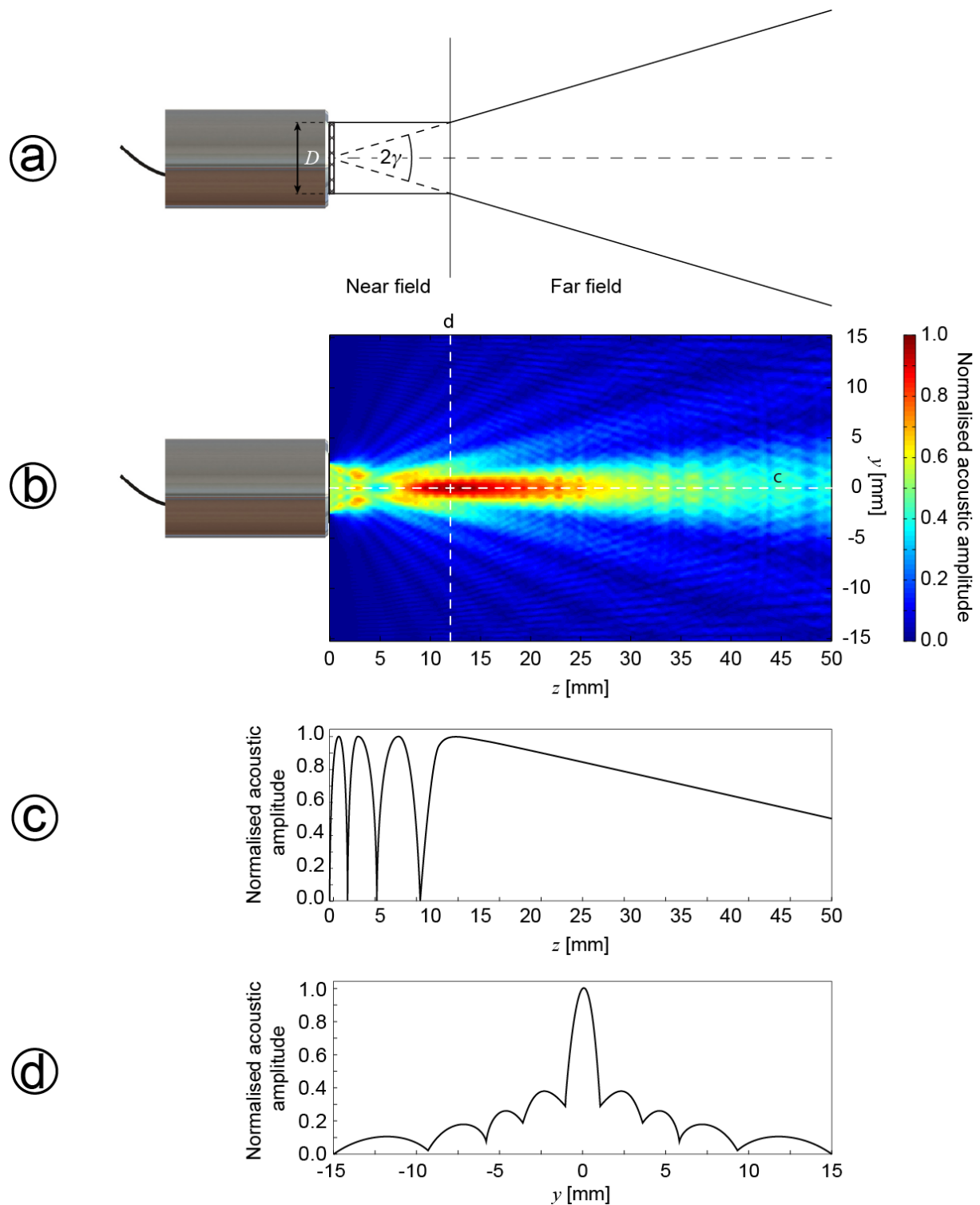
205

206 where D is the transducer diameter and N is the near-field length. In the far field, the
 207 sound field propagates with an opening angle 2γ :

208

$$\sin \gamma = 1.22 \frac{\lambda}{D}. \quad \text{Eq. 9}$$

209



210

211 **Figure 11 – Profiles of a sound field: (a) schematic; (b) 2D; (c) axial; (d) lateral.**

212

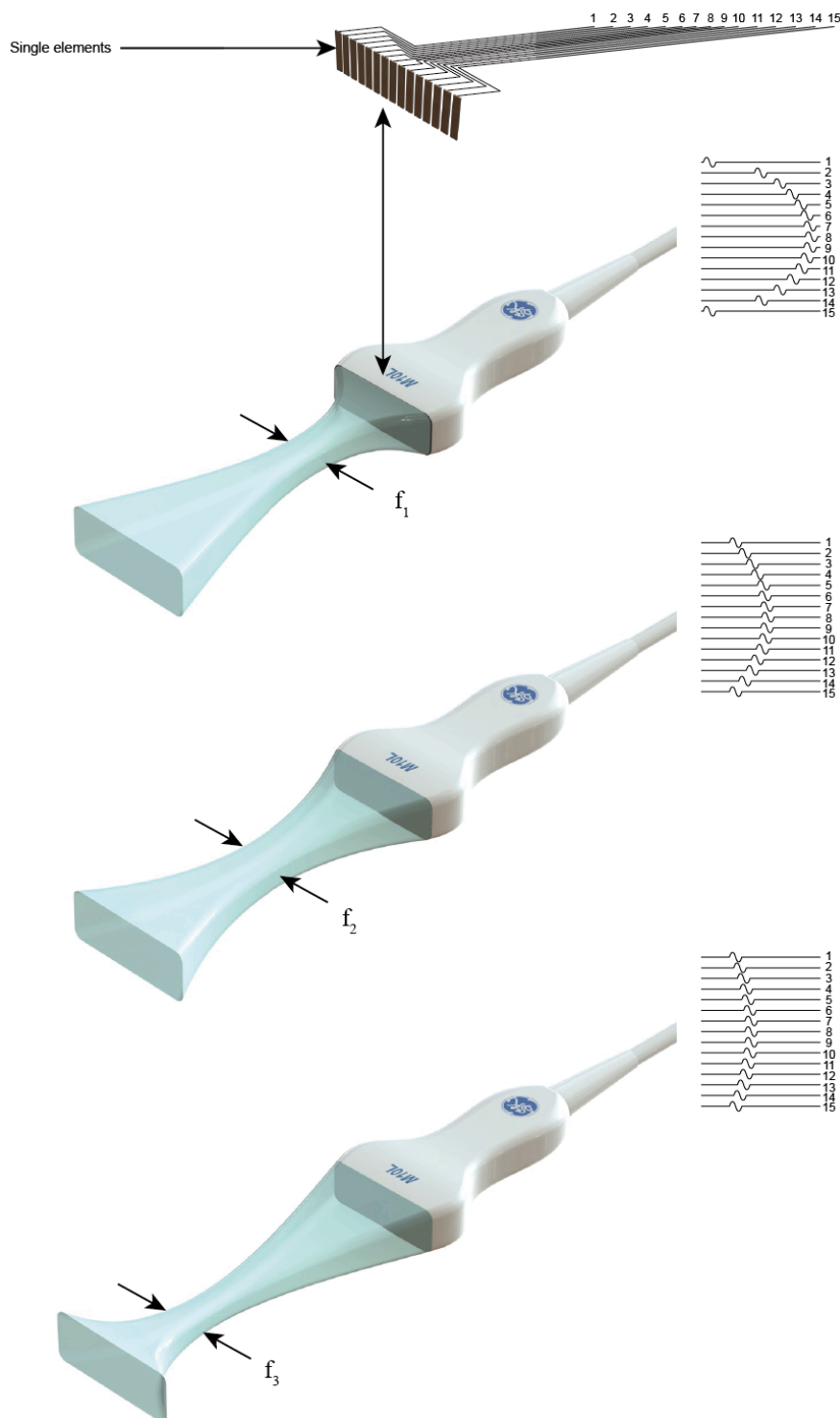
213 The axial plane separating near field from far field is the natural focus of a

214 transducer. A sound field can be further geometrically focussed by adding an

215 acoustic lens to the transducer surface.

216

217 In clinical ultrasound machines, multi-element transducers are used (cf. **Figure 12**).
 218 These consist of arrays of transducer elements lined up in the lateral direction, which
 219 can be individually controlled, allowing for variable beam focussing (cf. **Figure 12**).
 220 Such so-called phased arrays exist in numerous layouts, including curvilinear, 1.5D,
 221 and 2D variations (cf. Figure 13).
 222

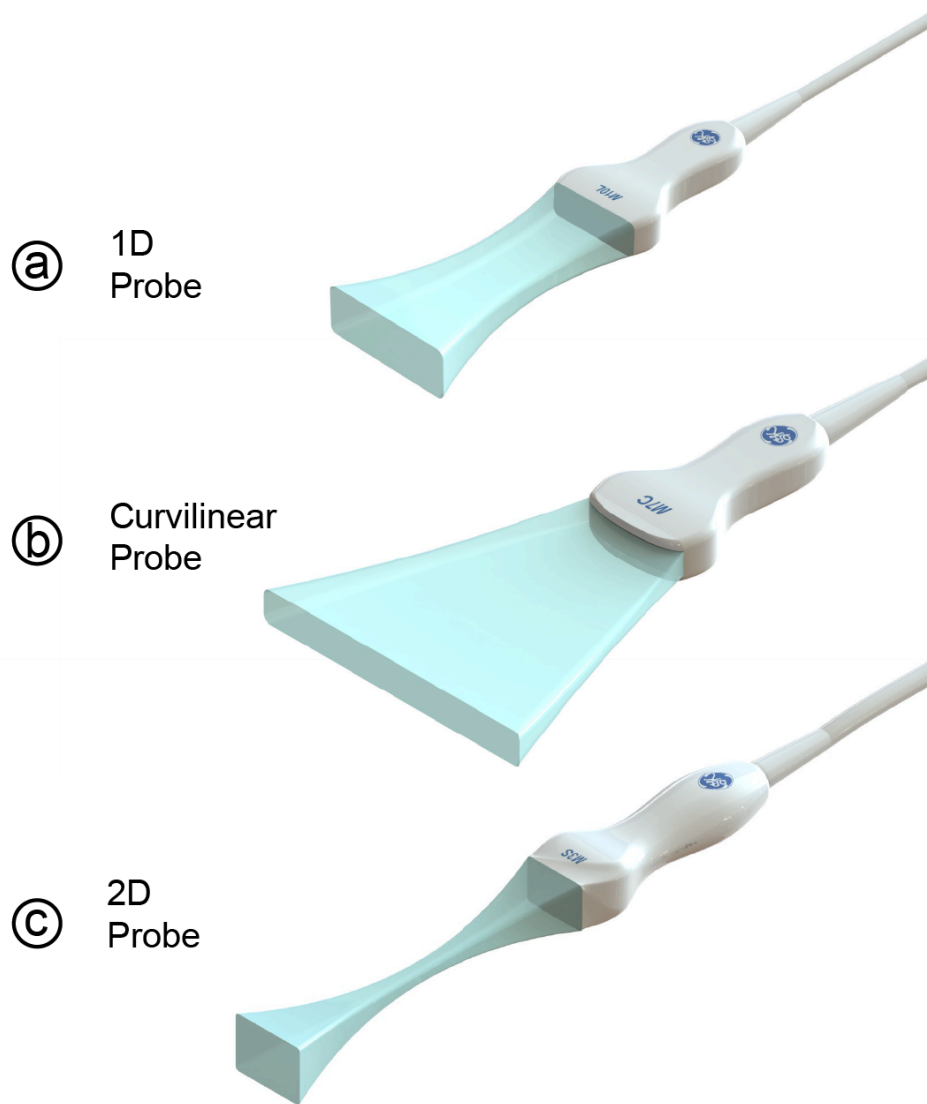


223

224

Figure 12 – Phased-array probes with variable focusing.

225



226

227 **Figure 13 – Phased arrays and their beam profiles: (a) 1D; (b) curvilinear; (c)**
 228 **2D.**

229 *Imaging*

230 Unlike a continuous sound wave, ultrasound for diagnostic imaging is transmitted as
 231 a pulse sequence (cf. Figure 14). After transmission of a pulse with a certain centre
 232 frequency, backscattered signal from tissue is received until the next pulse is
 233 transmitted. The pulse repetition frequency (PRF) is the number of pulses per time
 234 unit. The duty cycle is the percentage of transmission time, equal to the pulse length
 235 times the PRF. The theoretical maximum distance of imaging is

236

$$R_{\max} = \frac{c}{2 \text{ PRF}} \quad \text{Eq. 10}$$

237

238 Beware that the local speed of sound varies for different tissues. Therefore,
 239 ultrasound images built from the two-way travel times recorded are not converted to
 240 actual depth. The most commonly used mean speed of sound for quasi-depth
 241 conversion is 1540 m/s, which is the mean speed of sound in soft tissue (*cf.* Table 2).
 242 As this is just a chosen value, great care should be taken when drawing conclusions
 243 from quantitative spatial measurements using ultrasonic imaging.

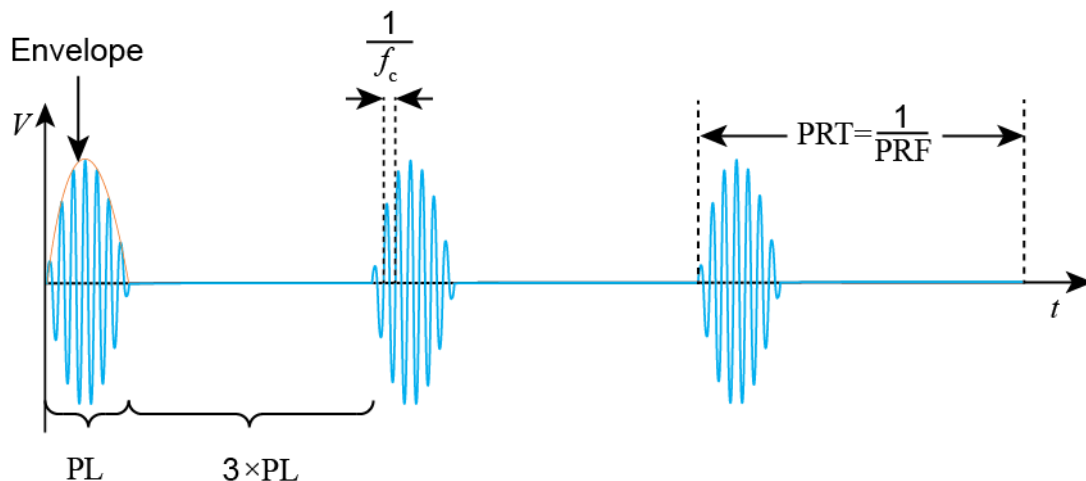
244

Material/tissue	c [m/s]
Air	330
Silicon oil	980
Water	1490
Blood	1570
Fat	1460
Muscle	1580
Bone	3500
Mean in soft tissue	1540

245

Table 2 – Speed of sound for different biomaterials.

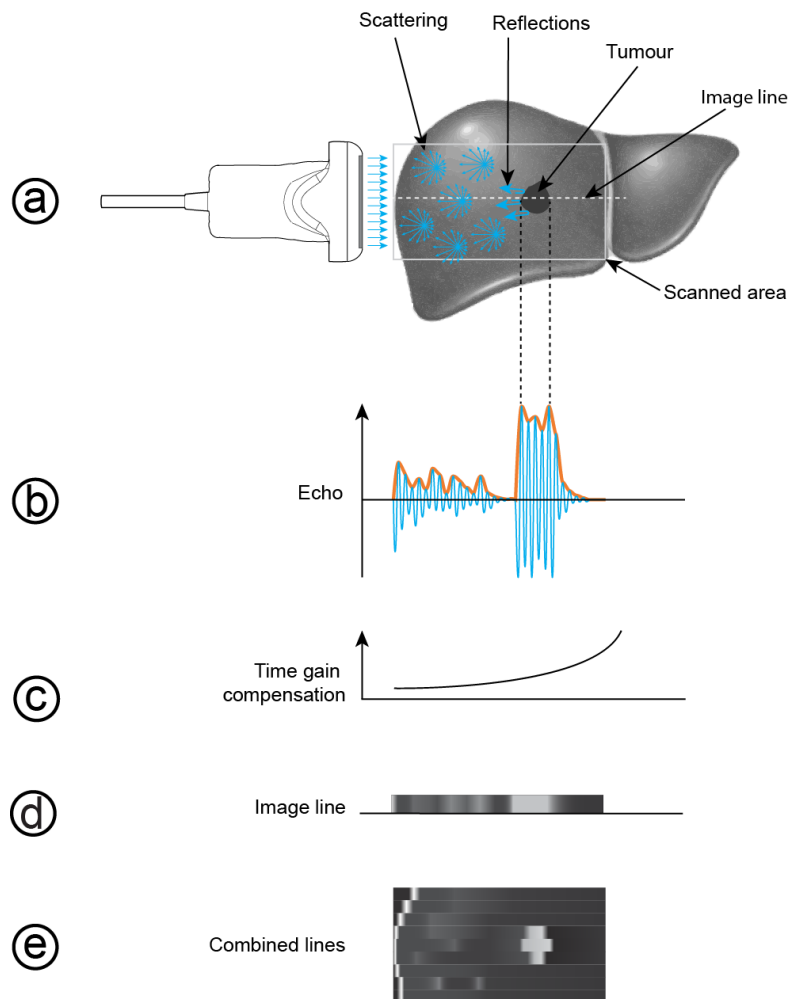
246



247

248 **Figure 14 – Pulsed transmit signal, with a centre frequency f_c , pulse length PL,**
 249 **pulse repetition time PRT, pulse repetition frequency PRF. The duty cycle of this signal**
 250 **is 25%.**

251



252

253

254

255 Figure 15 shows how a single beam is creating a line in an ultrasound image. If only
 256 the signal amplitude ("A") is considered, the imaging mode is called A-mode,
 257 whereas the amplitudes are represented by spots of brightness ("B") are more than
 258 one dimension, we speak of B-mode. Table 2 shows some typical applications of B-
 259 mode and the frequencies of choice. B-mode is the default clinical ultrasonic imaging
 260 method. B-mode is often combined with Doppler methods to track moving targets.

261 Using probes with mechanically moving transducers or 2D arrays, 3D B-mode scans
 262 can be recorded.

263

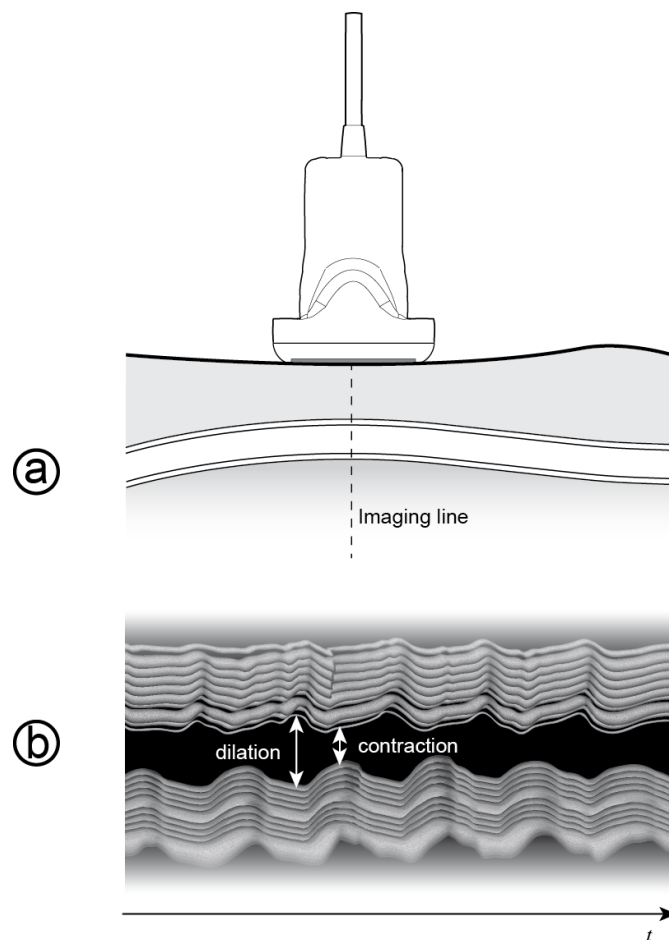
264

Figure 15 – Principle of B-mode imaging.

Frequency [MHz]	Penetration depth [cm]	Target organ
2-3	30	Deep abdomen
4-5	20	Adult heart
12-15	3	Mammae, Thyroid, Endosonography
20-50	1	Eye

265 **Table 3 – Some B-mode applications and their fundamental imaging**
 266 **frequencies.**

267



268

269

Figure 16 – M-mode imaging principle.

270 Displaying the received signal along one beam in motion (“M”), *i.e.*, as a function of
 271 time is called M-mode. The high time resolution allows for detailed study of
 272 periodically moving objects, such as the heart or a blood vessel (*cf.* Figure 16).

273

274 Resolution is the minimum distance between two points for them to be discriminated
 275 as separate points. The axial resolution is equal to the speed of sound divided
 276 through by twice the bandwidth. Hence, the wider the bandwidth (or the shorter the

277 pulse length) is, the smaller (better) the axial resolution is. The lateral resolution is
278 proportional to the wavelength of the sound and the transducer focal depth, and
279 inversely proportional to the transducer aperture.

280

281 **Safety indices**

282 The mechanical index (MI) gives an indication for the mechanical damage of tissue
283 due to inertial cavitation: the ultrasound-induced formation of transient cavities:

284

$$MI = \frac{PNP}{\sqrt{f_c}}, \quad \text{Eq. 11}$$

285

286 where PNP is the maximum value of the peak-negative pressure anywhere in the
287 ultrasound field (measured in water but corrected for a different attenuation)
288 normalised by 1 MPa and f_c is the centre transmit frequency normalised by 1 MHz.

289 At $MI < 0.3$, the acoustic amplitude is considered low enough for neonatal scans and
290 pregnant women. At $0.3 < MI < 0.7$, there is risk of minor damage to neonatal lung
291 and intestine. At $MI > 0.7$, there is a theoretical risk of inertial cavitation and a more
292 substantial risk if ultrasound contrast agents are being used.

293 Although the validity of the MI has been disputed, especially if an ultrasound contrast
294 is used, there is currently no alternative available to judge the safety from cavitation-
295 related damage in clinical settings.

296 Another, disputed, safety index is the thermal index (TI). It is a rough indicator of the
297 temperature rise in tissue during ultrasound exposure, and defined by the ratio of the
298 transmitted power and the estimated power needed to raise the tissue temperature
299 1°C . It should be noted that the TI does not indicate the actual temperature rise.
300 Based on thermal indices, limitations to ultrasound exposure times have been
301 recommended.

302 Near-future research will have to concentrate on redefining the safety standards.

303

304

305 **Further reading**

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 307 *Equipment*. Cambridge: Cambridge University Press 2010.
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 313 *Scattering: Scattering and Inverse Scattering in Pure and Applied Science*.
 314 London: Academic Press 2002:162-174.

315
316 **Appendix**

317 The equation of motion is

318

$$\frac{\partial p}{\partial x} = -\rho \frac{\partial^2 u}{\partial t^2}, \quad \text{Eq. A 1}$$

319 where ρ is the density of the medium.320 The linear 1-dimensional wave equation gives the sound pressure as a function of
321 space and time:

322

$$\frac{\partial^2 p}{\partial t^2} = \frac{\kappa}{\rho} \frac{\partial^2 p}{\partial x^2} = c^2 \frac{\partial^2 p}{\partial x^2}, \quad \text{Eq. A 2}$$

323 The speed of sound $c = \lambda f = \sqrt{\frac{\kappa}{\rho}}$ is a material property of the medium. Here, κ is the

324 bulk (incompressibility) modulus. Solutions of the wave equation have the form:

325

$$p(x, t) = p_A \sin 2\pi(ft - kx) = p_A \sin 2\pi \left(\frac{t}{T} - \frac{x}{\lambda} \right), \quad \text{Eq. A 3}$$

326 where $k = \frac{2\pi}{\lambda}$ is the wave number and p_A is the pressure amplitude.

327